In the Specification

Recharging an implantable medical device 20 generally begins with placing a recharging head 30 containing a primary recharging coil ~~34~~ 32 against the patient's skin near the proximal side of the medical device 20. Some recharges 28 have an antenna locator that indicates when the recharge head 30 is aligned closely enough with the implanted medical device 20 for adequate inductive charge coupling. The recharge power transfer signal is typically a frequency that will penetrate transcutaneous to the location of the implanted medical device 20 such as a frequency in the range from 5.0 KHz to 10 KHz. The power transfer signal is converted by the implantable medical device 20 into regulated DC power that is used to charge a rechargeable power source 34. Telemetry can also be conducted between the recharger 28 and the implanted medical device 20 during recharging. Telemetry can be used to aid in aligning the recharger 28 with the implanted medical device 20, and telemetry can be used to manage the recharging process. Telemetry is typically conducted at a frequency in the range from 150 KHz to 200 KHz using a medical device telemetry protocol. For telemetry, the recharger 28 and implanted medical device 20 typically have a separate telemetry coil. Although, the recharging coil can be multiplexed to also serve as a telemetry coil.

Please substitute the paragraph beginning page 6, line 14, with the following paragraph:

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The secondary recharging coil 68 is coupled to the electronics 40 and can also be coupled to the rechargeable power source 58 in addition to the electronics 40. In various embodiments the secondary recharging coil 68 can be located on the housing proximal face 76, inside the housing 66, and remotely away from the housing 66. The secondary recharging coil 68 has a

proximal side 82 implanted toward a patient's skin and a distal side 84 implanted toward a patient's internal organs. The secondary recharging coil 68 is manufactured from a material with electromagnetic properties such as copper wire, copper magnet wire, copper litz, woven wire, gold alloy or the like. The secondary recharging coil 68 can be manufactured from a wide variety of sizes such as wire diameters in the range from about 0.016 cm (34 AWG, American Wire Gauge) to about 00.40 cm (26 AWG), or any other suitable diameter. The secondary recharging coil 68 is coupled to the recharging feedthroughs 80 with an electrical connection 86. The electrical connection 86 is protected with a hermitic seal to prevent the electrical connection 86 from being exposed to biological tissue or fluids. The hermetic seal is a biocompatible material and can take many forms including potting material, polymer encapsulation, coil cover with polymer seal, or the like.

A1 cont

Please substitute the paragraph beginning on page 7, line 6, with the following paragraph:

The embodiment in FIG. 4a has a secondary recharging coil 68 carried on the proximal face 76 of the implantable medical device 20 with the magnetic shield 70 positioned between the secondary recharging coil 68 and the proximal face 76. The external secondary recharging coil 68 increases recharge efficiency because the secondary recharging coil 68 is located just under the surface of the skin to decrease coupling distance, and the magnetic shield 70 is position to both attract flux lines to the area of the secondary recharging coil 68 and reduce flux lines from coupling into the housing 66 to reduce eddy currents in the housing 66. The embodiment in FIG. 4b has an internal secondary recharging coil 68 with the magnetic shield 70 positioned between the internal secondary recharging coil 68 and the electronics 40. The internal secondary

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recharging 68 reduces manufacturing complexity and the magnetic shield 70 improves coupling and reduces eddy currents induced into the electronics 70. The embodiment in FIG. 4c has a remote secondary recharging coil 68 located away from the housing 66 with the magnetic shield 70 positioned on the distal side 84 of the secondary recharging coil 68. The remote secondary recharging coil 68 permits the clinician more positioning options while the magnetic shield 70 improves coupling.

Please substitute the paragraph beginning page 9, line 14, with the following paragraph:

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FIG. 9a shows a simulation test configuration with a magnetic shield 70 under a secondary recharging coil 68, and FIG. 9b shows a simulation test configuration with a magnetic shield 70 covering the medical device housing 66. FIGS. 9a and 9b are not to scale. Both simulation test configurations were done using two dimensional finite element analysis magnetic modeling software such as that available from MagSoft located in Troy, New York. Also both simulation test configurations used the following parameters. The primary recharging coil ~~34~~ 32 has 250 turns of 0.051 cm diameter (24 AWG) magnet wire with an outer diameter of 4.572 cm (1.8 inches) and an inner diameter of 2.019 cm (0.795 inches) with a Toroidal magnetic core in the center having an effective relative permeability μ_R of 10. The secondary recharging coil 68 has 200 turns of 0.025 cm diameter (30 AWG) magnet wire forming a coil with an outer diameter of 3.302 cm (1.30 inches) and an inner diameter of 0.635 cm (0.25 inch). The medical device housing 66 is titanium having a thickness of 0.030 cm (0.012 inch). The separation between the primary recharging coil ~~34~~ 32 and the secondary recharging coil 68 is 1.0 cm (0.394 inch). The recharge power transfer signal is 150 VAC peak-to-peak at 8.0 KHz. The magnetic

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shield 70 in FIG. 9a is composed of alternating 0.002 cm (,001 inch) thick layers of Metglass and air gap with the secondary recharging coil 68 located 0.013 cm (0.005 inch) above the magnetic shield 70. The magnetic shield 70 in FIG. 9b as the magnetic shield 70 described for FIG. 9a and in addition a similar magnetic shield 70 covering the medical device 20 sides and bottom.

Please substitute the paragraph beginning on page 10, line 13, with the following paragraph:

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FIG. 10b shows simulation results with a magnetic shield 70 placed under the secondary recharging coil 68 and power transfer signal flux lines 96 interacting with the secondary recharging coil 68 and a medical device housing 66. Power loss in the medical device housing 66 is 0.143 Watts and the coupling efficiency is 25.4%. The simulation results show improved recharging efficiency through enhanced electromagnetic coupling between the secondary recharging coil 68 and a primary recharging coil ~~34~~ 32. The improved electromagnetic coupling between the primary recharging coil ~~34~~ 32 can be in the range from about 10% to 28% coupling efficiency at about one centimeter. Electromagnetic coupling efficiency is calculated with the following equation: $\textit{Coupling Efficiency} = \frac{Pout}{Pin} \times 100\%$ where Pout is measured at the secondary recharging coil 68 and *Pin* is measured at the primary recharging coil ~~34~~ 32. The recharging efficiency is also improved through reduced eddy currents in the housing 66. Reducing eddy currents during recharging also reduces medical device 22 temperature rise during recharging for improved safety.

Please substitute the paragraph beginning on page 11, line 9, with the following paragraph:

FIG. 11 shows a method for enhancing electromagnetic coupling of an implantable medical device external recharging coil embodiment. Positioning a secondary recharging coil 98 in operational relationship to an implantable medical device 20. Positioning a magnetic shield 100 on the distal side of the secondary recharging coil 68. Attracting electromagnetic flux lines 102 from a primary recharging coil ~~34~~ 32 to the secondary recharging coil 68 with the magnetic shield 70. Improving electromagnetic coupling between a primary recharging coil ~~34~~ 32 and a secondary recharging coil 68. The improved electromagnetic coupling 104 between a primary recharging coil ~~34~~ 32 and the secondary recharging coil 68 is in the range from about 10% to 28% coupling efficiency at about one centimeter. Improving efficiency 106 of energy transfer from the primary recharging coil ~~34~~ 32 to the secondary recharging coil 68. The efficiency of energy transfer is improved because less energy is lost to eddy currents in the housing 66.